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NUMERICAL INVESTIGATION OF BLOOD FLOW THROUGH ABDOMINAL ANEURYSMS

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ABSTRACT

Cardiovascular diseases are very common in today's world. It is of great importance to simulate the behavior of arteries subjected to various anomalies using computational methods, as it will help the clinicians in the early diagnoses of the disease. This paper enumerates the use of computational fluid dynamics (CFD), for simulating blood flow phenomenon in human arteries, especially Abdominal Aortic Aneurysm (AAA). The use of the CFD in simulating the anatomically realistic models is discussed. Different methodologies applied for assessing the effectiveness in predicting the behavior of blood flow in arteries is presented. In this study, it was observed that higher wall shear stress is observed at peak systole. It reduces at early diastole which leads to flow separation and recirculation inside the aneurysm dome.

KEYWORDS: Abdominal Aortic Aneurysm, Computational Fluid Dynamics & Velocity Vectors

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INTRODUCTION

The vascular system in humans comprises of the organs, which transport blood throughout the body. The blood circulation is divided into two parts and forms a closed loop. The systemic circulation originates from the left chamber of the heart, from where the blood flows into the aorta. The pulmonary circulation begins in the right ventricle and ends in the left atrium. Cardiovascular hemodynamics refers to the study of blood circulation to the heart and in turn blood circulation regulated by the heart. Cardiovascular disease (CVD) is a general term for conditions that affect the heart or blood vessels. One such CVD is abdominal aortic aneurysm (AAA). An aortic aneurysm is a condition in which the aorta expands to 1.5 times or more its normal size. There is usually no symptom except when it gets ruptured. The most common occurrence of such kind of aneurysm is in abdominal aorta, but in some cases it can also be found in thoracic aorta. Aortic aneurysms weaken the wall of the aorta and increase the risk of aortic rupture. Rupture of such aneurysms is followed by internal bleeding, and can be fatal unless it is treated immediately. Abdominal aortic aneurysms (AAAs) are more common than their thoracic counterpart. One reason for this is that elastin, the principal protein that acts as a load-bearing entity present in the wall of the aorta, is reduced in the abdominal aorta as compared to the thoracic aorta. The risk of rupture of an AAA is related to its diameter and shape. Long aneurysms are considered less rupture prone than shorter aneurysms.

Blood flow in arteries is controlled by unsteady flow phenomena. The pulsatile nature of the arterial blood flow is its most important characteristic. Arterial flow under normal circumstances is considered to be a laminar flow. In certain circumstances, unusual hemodynamic conditions create an abnormal biological response. Changes

in artery geometry such as the development of stenosis, implementation of catheter devices and stiffening of arterial walls causes turbulence and reduces the required blood flow [1][2]. To capture the flow characteristics of the blood for patient specific models, different boundary conditions are implemented at the artery inlet and outlet depending on the anatomic location, age of the patient, nature of blood flow in that area and other decisive factors. Different boundary conditions yield different end results and hence a comprehensive study of these is of utmost importance. Generally, a velocity/flow profile is prescribed to the inlet and pressure plot is defined for artery outlet to model the response of back pressure due to stenosis or other constricting phenomenon. The geometry of the artery is modeled each time from beginning for the patient specific cases, and accordingly material properties and other structural and fluid boundary conditions are provided.

The pressure exerted by the flowing blood upon the walls of arteries is called blood pressure. The pumping action of human heart is responsible for the development of pressure waves. In many cases, due to the presence of many abnormalities, the blood pressure tends to rise. Hence, a very important role is played by the simulation studies of pressure waves in the human system in diagnosing the affected area of the human body. A theoretical analysis of the pressure wave propagation through a tube containing viscous and incompressible fluid with initially pre-stressed condition was considered [1]. The fluid was Newtonian and the tube was considered to be elastic and isotropic. It was found that, for large pressure rise, the longitudinal displacements predicted by the theory are very huge compared to the observed longitudinal oscillations of arterial walls. Therefore, it was necessary to introduce some sort of longitudinal constraint, so as to use the theory in the study of blood flow in arteries. The pressure wave propagation through a viscous fluid flowing in an elastic thick walled orthotropic tube was studied in [2], the fluid flow was considered with low Reynolds number laminar flow and the presence of reflected waves. The authors observed a decrease in the flow of blood as the pulse propagated away from the heart. Cox et al. [3] considered the propagation of wave in a thick-walled incompressible viscoelastic tube. The motion of the wall was explained by classical theory of elasticity. From their results, it was found that the fluid impedance predicted by this model was smaller than the rigid tube model values. Wang and Tarbell [4] studied the non-linear behavior of Newtonian fluid subjected to the oscillatory pressure gradient. They assumed the blood to be a homogenous and incompressible fluid. The artery was assumed to be an isotropic, thin-walled, and elastic tube with negligible longitudinal wall motion. From their study, it was found that, under the condition of no mean flow in the arteries, the nonlinear convective acceleration is generated a finite mean pressure gradient. The Wave propagation across an elastic, tapered, thin walled long circular tube was modelled by taking the value of mean pressure as 100 mmHg [5]. The authors assumed that, a pre-static deformation of the arteries and an additional dynamic deformation would occur when the blood flows through the arteries. The equations were developed for static deformation and modified to incorporate the dynamic deformation concept assuming the fluid to be incompressible, non-viscous and one dimensional. According to their research, wave speed increased with a decrease in tube radius. Matthys et al., [6] developed an experimental set up to study the wave propagation in arterial systems. their studies concluded that one dimensional models were able to capture all the wave propagation features. The importance of wall visco-elasticity was also discussed. Young Woo Kim et al [7] analyzed the relationship between wall elasticity and blood flow wave reflection. A Non-Newtonian, viscoelastic model of the blood was assumed. Impedance boundary condition was given to the outlet of computational domain to develop the required effect of wave reflection. The difference in flow rate and shear stresses were analyzed for a wide range of ages. According to their studies, it was found that curved points had higher shear rate. The risk of stenosis could be predicted precisely if the relationship between stenosis and wall shear stress is defined. It was also found that wave reflection was faster in the vessel of older person than compared to that of a younger person.

There are various material models that can be used for artery walls to capture the wall deformation accurately. The most common of them are rigid artery models, elastic material models (Hooke's law) and hyper elastic material models. The two types of hyper elastic material models define the strain energy function in different ways. The phenomenological model treats the problem from the viewpoint of continuum mechanics whereas the second model considers the material response from the viewpoint of microstructural analysis. In [8] a two layered structural model was developed for the study of arteries. By taking the media and adventitia layers as fiber reinforced composite layers, it was concluded that this model could be successfully used to study the wall deformation in arteries. In [9] a two-layer idealized model of artery was proposed. The outer layer of the artery was modelled using the structural model as proposed by Holzapfel in [8]. The inner layer was assumed to be made of smooth muscle tissues. The material parameters such as shape, volume fraction and orientation for this model were determined by using microscopic measurements. The model was loaded with inner pressure to simulate the mechanical response of an arterial segment during inflation. Comparison of the predicted results of the model with the mechanical response of real arterial segment was performed. The theoretical curves that were plotted, exhibited a good agreement with experimental data. The influence of non-uniform wall elasticity on the change of shape of an artery throughout steady and pulsatile blood flow was investigated in [10]. The authors recreated the artery geometry using the data derived from a case for a stenosed left coronary artery and the blood flow in the artery was modeled using power-law fluid. The FSI effect on pathological (stenosis) obstacles can be studied through 3 physical parameters: minimum velocity, maximum velocity and the disturbance length [11].

The mechanical interaction between blood and arterial wall is mainly responsible for the propagation of pressure wave from the heart to whole body. This interaction is the main factor, which help to regulate blood pressure in the body. In FSI the computational stability is depends on the coupling. A good computational stability can be obtained in Time periodic coupling [12]. The computational cost and effort can be reduced by considering the arterial wall as linearly viscoelastic in a fully coupled fluid and structure interaction. [13] Hemodynamics in large arteries can be analyzed using finite element base algorithms also, the wall thickness plays an important role in FSI. The uniform wall thickness could underestimate risk in aneurysms while the rigid wall will overestimate the wall shear stress [14] [15]. Toloui et al., [16] examined the wall deformability and blood rheological properties in the wall shear stress. In computational analysis the arteries can be modeled as two layered structural model [8][9] and the blood flow can be analyzed for steady and pulsatile flow conditions [10] [11] [17]. The blood flow model is generally decided by the Reynolds number [18] [19]. For the large size arteries, the Newtonian blood flow model will capture flow characteristics with the acceptable limits.

The need to understand this flow of blood through the circulatory system has inspired a number of research activities in hemodynamics. A vital part is played by these simulation studies of flow waves in the human system to identify the affected area of the human body and finding new solutions to various diseases related to circulatory system. Hemodynamic aspects, such as wall shear stress (WSS) and oscillatory shear index (OSI) have been associated to origination and development of plaque in arteries. Because of the restrictions of experimental facilities, numerical simulation plays an essential role in the analysis of the flow. Various computational approaches have emerged as influential tools for modeling blood flow and vessel deformation in the vascular system. Fluid-structure interaction (FSI), a computational method, has emerged as of great assistance to model blood flow through arteries while monitoring appropriate structural characteristics. In this report, we deal with the simulation of fluid-structure interface (FSI) problems in hemodynamics.

METHODOLOGY

A patient specific MR imaging based 3-D model of abdominal aorta is created with bifurcation. The aorta is considered to have aneurysm. A 3D anatomical reconstruction software was used to convert them into a realistic 3-D surface model as shown in Figure 1. Operations like segmenting, smoothening the artery surface, trimming the artery geometry and making it hollow are performed on the same as shown in figure 2.

To capture the flow characteristics of the blood in the arterial bifurcation, a time-varying velocity profile has been given to the artery inlet (figure 3) and a constant pressure boundary condition is implemented at both artery outlets. The value of constant pressure at artery outlet is 530 Pa. A transient structural is performed by using ASYS transient Structural and fluid flow is analyzed by using ANSYS CFX.

The final product exported out of mimics is a surface file having stereolithographic format. CATIA V5 surface reconstruction module has been used to convert the surface models into solid bodies with volumes as shown in figure 4 and 5. Solid and the fluid bodies are then assembled as shown in figure 6.

The final geometry has many faces on its surface; hence, a virtual topology is required for the ease of meshing (figure 7 and figure 8). The virtual topology merges the multiple faces into a single face and makes it easier for the software to read the geometry during mesh creation and setting up boundary conditions. This is done using ANSYS Mechanical.

The final geometry after the creation of virtual topology has been shown in Figure 9. Further information related to material model, mesh and setup of structural part is given in table 2. Figure 8 shows the structural geometry after mesh and Figure 9 shows the fluid geometry after mesh. To obtain a better aspect ratio for the mesh, both the geometries have been sliced into 2 parts each. Further information related to material model, mesh and setup of fluid part is given in Table 3. The solution was obtained for a total of three cardiac cycles and the results have been analyzed for third cycle.

The geometric detail of the constructed aortic model is detailed in Table 1.

Table 1: Geometric Information

Parameters	Current Simulation
Wall thickness	2.5 mm
Maximum transverse diameter	56 mm
Length	132 mm
Inlet diameter	28 mm
Outlet diameter 1	18 mm
Outlet diameter 2	20 mm
Neck diameter	31 mm

Table 2: Structural Information

No. of Elements	No. of Nodes	Mesh Method	T (s)	Time Step(s)	No. of Pulse Cycles	E (dyne/cm2)	μ
16988	34026	Tetrahedron	3.0	0.02	3	4.07E+6	0.45

Table 3: Fluid Information

No. of Elements	No. of Nodes	Mesh Method	T (s)	Time Step (s)	No. of Pulse Cycles	Rho (g/cm3)	μ (dyn/cm2)
48477	70123	Tetrahedron	3.0	0.02	3	1.06	0.04

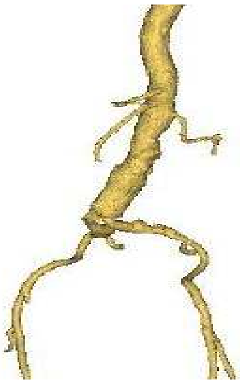

Figure 1: 3D Model of Abdominal Aorta.

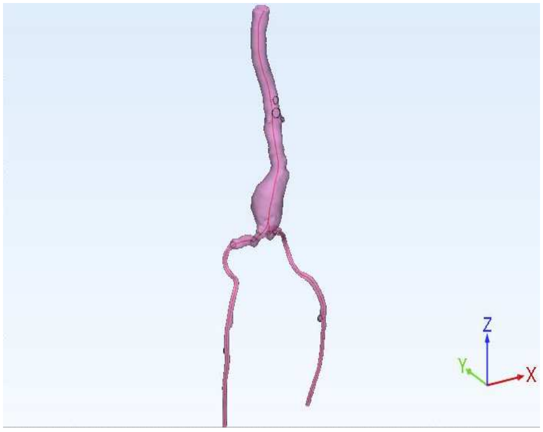

Figure 2: 3D Reconstruction of the AAA Model.

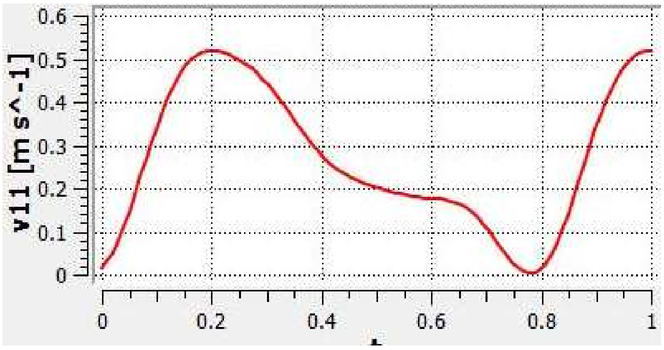

Figure 3: Inlet Velocity Boundary Conditions.

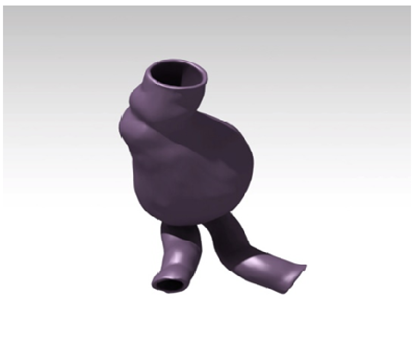

Figure 4: Arterial Geometry. **Figure 5: Fluid Domain.**

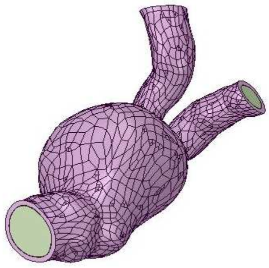

Figure 6: Final Assembly of Fluid and Solid Domain.

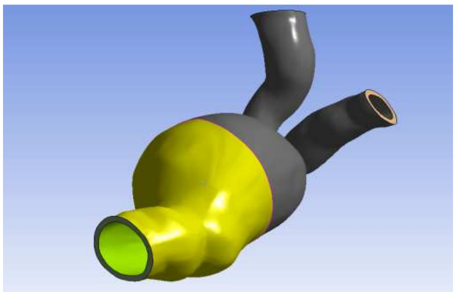

Figure 7: Solid Body Virtual Topology.

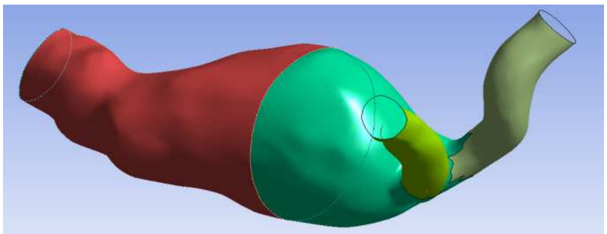

Figure 8: Fluid Body Virtual Topology.

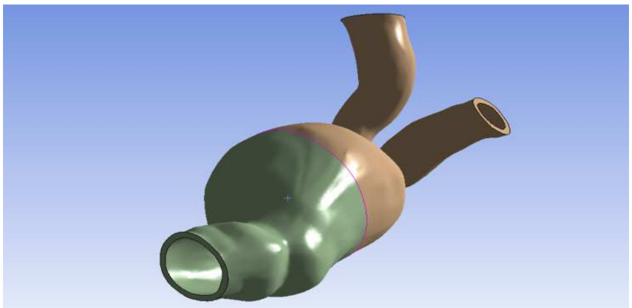

Figure 9: Final Geometry.

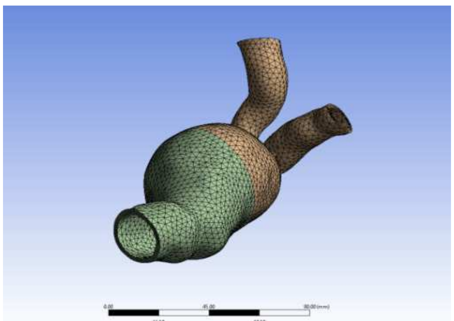

Figure 10: Structural Mesh.

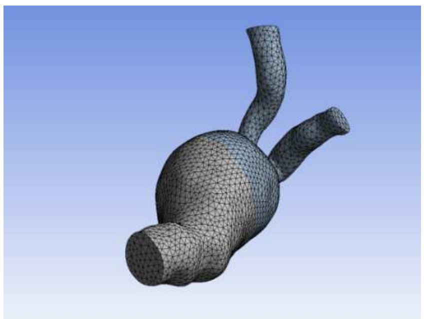

Figure 11: Fluid Mesh.

RESULTS AND DISCUSSIONS

A time dependent analysis to determine the effects of blood flow on aneurysm was conducted for Abdominal Aorta. Aneurysm transverse diameter, shape and parent artery diameter are the predominant factors known to influence the nature of blood flow within an aneurysm. The solutions were obtained for three cardiac cycles 0.8 seconds each long, and the results were analyzed for the third cycle. The third cycle begins at 1.58s and the peak systole occurs at 1.8s. The cycle ends at 2.4s. The AAA demonstrates complex flow patterns over the course of the complete cardiac cycle. Two planes were created at strategic locations of maximum aneurysm sac diameter (S1) and aneurysm neck (S2). Pressure and flow variations were calculated for the two planes. Total mesh displacement was also calculated for the solid and the fluid domains. A comparison graph is shown in figure 12.

At peak flow (1.8s), a characteristic attached flow pattern is observed throughout the aneurysm with nearly stationary flow pattern. During systolic deceleration, a flow separation is observed and the vortex begins to generate and propagate in the aneurysm sac. The results have been shown for the peak systole (0.18s) and systolic deceleration point (2.12s), where the flow begins to separate. Figure 13 and 14 shows the preure contour and velocity vector at S1, respectively.

During this time-period, the flow is attached to the wall and separation has still not begun. It can be seen in the velocity streamlines (figure 16). The value of maximum velocity at the location of maximum diameter at peak systole is found to be 0.237 m/s. The maximum velocity at aneurysm neck is 0.565 m/s.

This stage is depicted with significant and asymmetric flow recirculation near the aneurysm neck. As a result of the flow separation due to reduction in velocity, vortex is created inside the aneurysm (figure 19). During systolic deceleration and early diastole a large recirculating flow region fills the aneurysm sac.

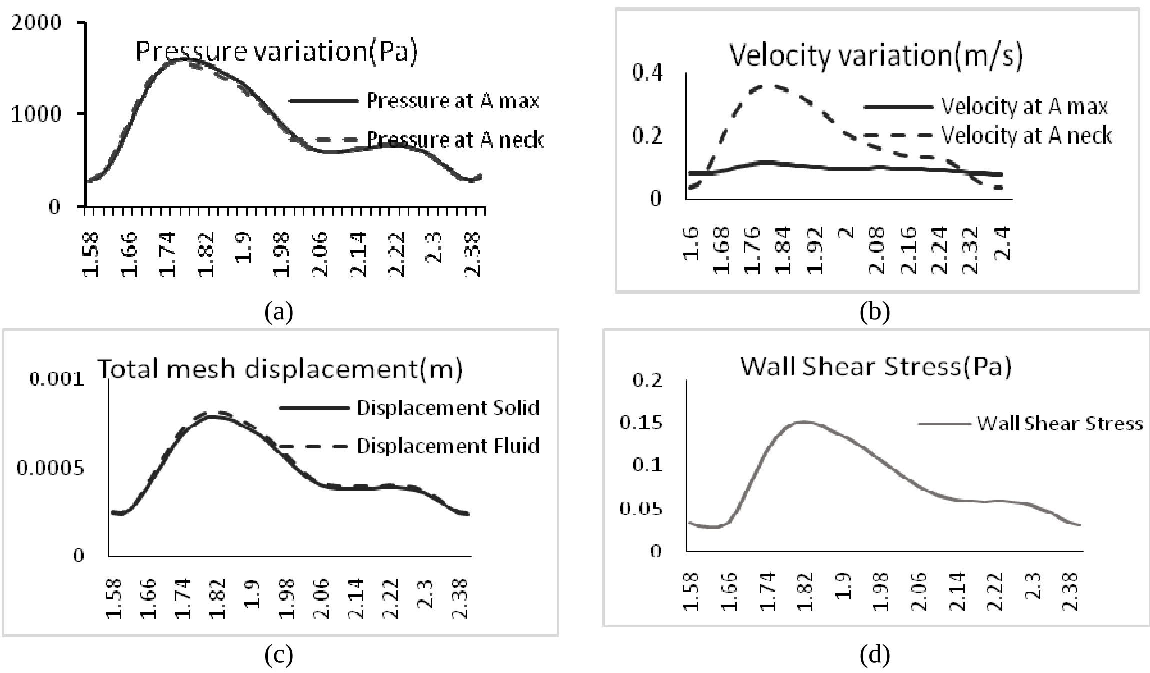

Figure 12: Comparison Graph. (a) Pressure Variation. (b) Velocity Variation (c) Total Mesh Displacement (d) Wall Shear Stress.

Figure 13: Pressure Contour.

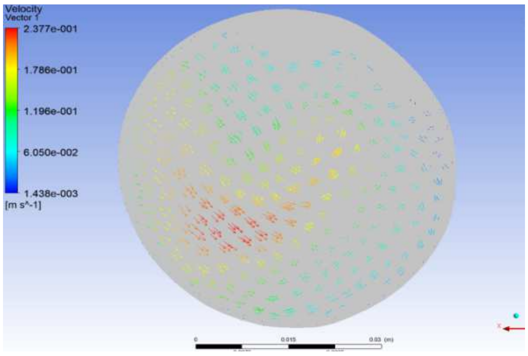

Figure 14: Velocity Vector at S1.

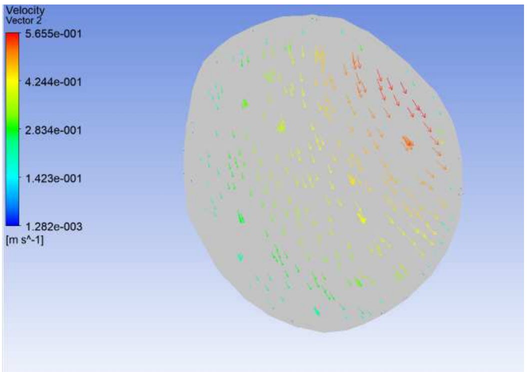

Figure 15: Velocity Vector at S2.

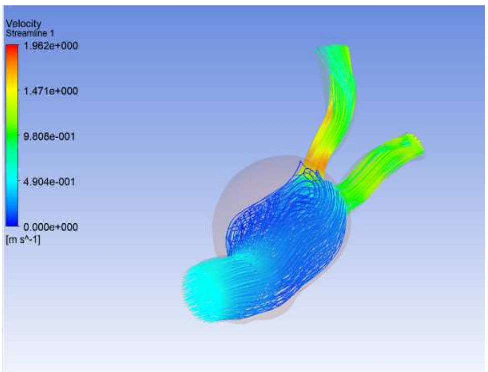

Figure 16: Streamlines at Peak Systole.

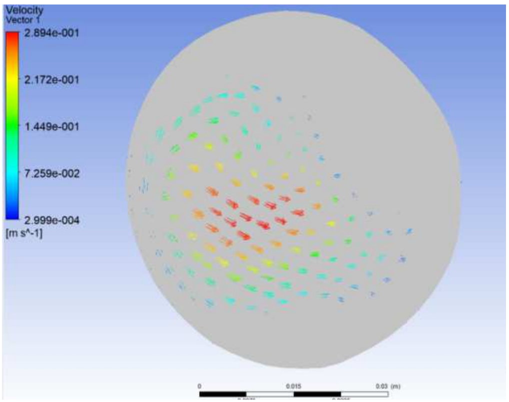

Figure 17: Velocity Vector at S1 at Diastole.

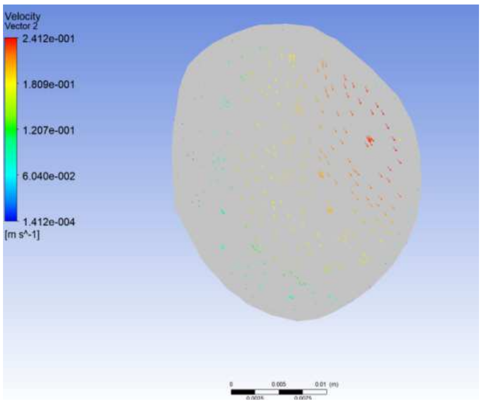

Figure 18: Velocity Vector at S2 at Diastole.

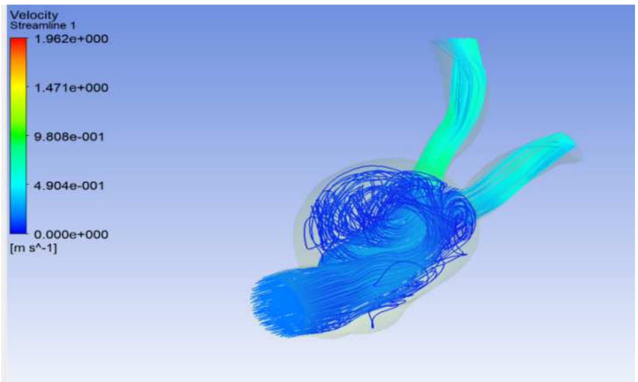

Figure 19: Velocity Streamlines at Diastole.

CONCLUSIONS

In this work, Fluid-Solid Interactions (FSI) is applied to a patient-specific clinically-relevant blood flow problem through abdominal aneurysms. The use of image based modelling of realistic geometry of arteries using CT MRI and CFD studies of blood is extensively used to study the hemodynamics of blood flow. It was found that use of realistic models helps in better understanding of hemodynamics when compared to idealized models. Although MRI gives hemodynamic parameters based on geometry of the lumen, it was found that CFD results are more reliable than the medical imaging.

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